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In-plane-gate a-IGZO thin-film transistor for high-sensitivity pH sensor applications

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ABSTRACT

In this paper, we propose a thin-film transistor (TFT) with an in-plane-gate (IPG) structure, in which the channel and bottom-gate electrodes are located on the same plane to enhance the sensitivity of the pH sensor. The transistor was applied to an extended-gate field-effect transistor (EGFET) system for disposable sensor implementation. The fabricated IPG pH sensor exhibited amplified pH sensitivity owing to the improved capacitive coupling effects. Moreover, we measured the dependence of the pH sensitivity on the IPG area and evaluated the performance of the pH sensor with the IPG structure. The fabricated IPG pH sensor showed a linearity of 99% or more even for a small IPG area, and the amplification of the sensitivity was improved with further reduction in the area. Notably, we achieved a high sensitivity of 387.58 mV/pH, which is significantly greater than the Nernst limit (59 mV/pH), at the lowest IPG area. To evaluate the stability and reliability of the fabricated IPG pH sensor, non-ideal effects, such as hysteresis and drift characteristics, were considered. The IPG structure exhibited enhanced stability and reliability. Accordingly, we believe that EG-type biosensors with an IPG structure can be useful for biomedicine, clinical diagnosis, environmental monitoring, and point-of-care-testing (POCT) systems.

1. Introduction

As the interest in early diagnosis of various diseases and quality of life increases, biosensors have received considerable attention over the last several decades. A biosensor is an analytical device used to detect an analyte, wherein a biological component is combined with a physicochemical detector. Several biosensors have been developed, wherein chemiluminescence, quartz crystal microbalance (QCM), and surface plasmon resonance (SPR) methods have been used to detect biological elements; however, these methods are unsuitable for home medical care. Chromatography cannot be used to measure bio-materials in real-time, and SPR and chemical luminescence equipment are too large and heavy for home use [1]. As a substitute for this, a potentiometric biosensor method based on field-effect transistors (FETs) has emerged. As FET-based biosensors have high sensitivity, similar to those of SPR and QCM, they are a promising and viable point-of-care diagnostic device for the rapid and sensitive detection and analysis of disease-related biomarkers. In particular, FET-based ion-sensitive sensors (ISFETs) are more advantageous as they can be used to selectively detect biological analytes in a real-time, label-free, and regentless manner [2,3]. In addition, they are inexpensive and highly compatible with complementary metal-oxide semiconductor (MOS) processes. The ISFET structure is similar to that of a metal-oxide-semiconductor field-effect transistor (MOSFET), and the potential is measured at the oxide–electrolyte interface [4,5]. However, existing ISFETs have disadvantages such as the lack of implementation of a disposable sensor due to weak chemical stability; sensitivity change due to optical elements; low isolation; and combined structure (of sensing membrane (detector) and transistor (transducer)). In contrast, an extended-gate FET (EGFET), which is a modified version of the ISFET, has a structure in which the EG; which serves as the detector for detecting the biomaterials, and the transducer, which is used to convert the potential change into an electrical signal; are separated [6,7]. Therefore, making the EGFET more suitable for disposable sensor implementation than ISFET. Moreover, it is confirmed that EGFET have detecting capability for real-biomaterials, such as drosophila LUSH odorant-binding protein [8], salivary sugar related with alzheimer's disease [9], and hepatitis B surface antigen [10]. However, the fundamental problem with the ISFET and EGFET is that the sensitivity of the sensors depends on the ability of the sensing membrane to have a Nernstian limit of 59 mV/pH at room temperature [11,12]. It should be noted that because the actual biological elements have a low signal, a high sensitivity is required to obtain sufficient signal margins compared to noise [1]. To overcome this major problem, in our earlier investigation on SOI MOSFETs and amorphous indium-gallium-zinc oxide (a-IGZO) thin-film transistors (TFTs), we proposed a pH sensor with a sensitivity higher than the

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Nernstian limit using the capacitive coupling phenomena in consideration of the specificity of the double-gate structure [13–16].

In this study, to more effectively amplify the sensitivity, which is generated by the capacitive coupling effect, we propose a novel TFT with an in-plane-gate (IPG) structure [17–20]. Because of the favorable operating characteristics of the IPG structure, it has been studied and employed for developing various promising devices such as logic devices [21,22], rectifiers [23], and neuron TFT [21,24]. On the other hand, the structural peculiarity of the IPG is more beneficial for improving the sensor sensitivity characteristics of the ISFET, compared to that in previously applied fields [25]. In the proposed device, the IPG is located on the same plane as that of the a-IGZO channel, and the capacitive coupling with the channel and the upper-gate (UG) was performed through the buried conducting layer. Here, the IPG and UG correspond to bottom and top gates, respectively, of the conventional double-gate structure transistors. The IPG TFT, which serves as a transducer, comprises an a-IGZO channel, UG SiO_2/IPG SiO_2 layers, an indium tin oxide (ITO) buried layer, source/drain (S/D) electrode, and UG and IPG electrodes. The EG, which serves as a detector, comprises a tin dioxide (SnO_2) sensing membrane having good sensing characteristics [13–16,26]. The microwave irradiation method has been applied to improve the electrical characteristics and instability of the a-IGZO channel [27]. As a result, the sensitivity of the IPG structure is improved. In addition, the smaller the area of the IPG, the greater is the sensor performance; the stability and reliability have been also improved owing to the structural specificity of the IPG structure.

2. Material and methods

First, we performed standard RCA cleaning on a p-type (100) 1–10 Ω·cm Si wafer. A 50-nm thick indium-tin oxide (ITO) film was used as the floating layer; this film is used to transfer the potential of the IPG for a-IGZO channel deposition using an RF magnetron sputter (20 sccm, 3 mTorr), thereby forming a buried conducting layer [17,19–21]. Thereafter, a 270-nm thick silicon dioxide (SiO_2) film was deposited with IPG oxide using an RF magnetron sputter. Accordingly, the distance from floating layer to IPG and IGZO channel is 270-nm thick. A 20-nm thick a-IGZO film was deposited using RF magnetron sputtering (30 sccm, 6 mTorr), and channel pattern was formed by photolithography and a 30:1 buffered oxide etchant (BOE, 30 volumes of 40% ammonium fluoride to 1 vol of 49% hydrofluoric acid). The patterns of the IPG and S/D electrodes were simultaneously defined using photolithography, and IPG and S/D electrodes of 100-nm-thick ITO were formed using a lift-off process. A 50-nm thick SiO_2 film was deposited using RF magnetron sputtering with the UG oxide of the a-IGZO TFT, and a 150-nm thick ITO UG electrode was formed. For the measurement of the fabricated device, S/D and IPG contact holes were formed using the 30:1 BOE etchant. Finally, to improve the electrical characteristics of the TFT, annealing was performed using the microwave irradiation method under a condition of 1000 W for 2 min in air ambient. The defined channel width and length of the fabricated IPG TFT are 20 and 10 μm, respectively, and the IPG area was in the range of 1200–32,000 μm^2.

For the implementation of disposable and inexpensive biosensors, the introduced IPG TFT has two structural advantages. The first is that the IPG does not require a contact-hole process for the bottom-gate contact, as in the conventional double-gate structures. Therefore, the process cost is reduced because the fabrication process is simplified. As the IPG is located on the same plane as that of the source/drain electrodes, it can be formed simultaneously with the source/drain electrode formation. Moreover, there are no stringent requirements on the type of material used for the bottom-gate oxide. For example, low-k material, organic material, or polymer material, which are otherwise difficult to be applied to contact-hole processes, can be employed in this method [19]. The other advantage is that the IPG enables the use of cheap and flexible substrates such as paper [18,19] and plastic [20].

To implement transducer, before thermally growing 100-nm thick SiO_2, a standard RCA cleaning was performed on the p-type (100) 1–10 Ω·cm Si wafer. To transfer the signal potential to the transducer, a 150-nm thick ITO film was deposited using RF magnetron sputtering. Thereafter, a 50-nm thick SnO_2 film was deposited as the ion-sensitive membrane using the RF magnetron sputtering. The RF power, chamber pressure, and Ar gas flow rate were 50 W, 3 mTorr, and 20 sccm, respectively. Finally, a polydimethylsiloxane (PDMS) reservoir was attached onto the SnO_2 film using silicon glue to inject the pH buffer solution.

We measured the pH sensitivity of the IPG TFT sensor using the following two methods. Single-gate (SG) mode sensing is a measurement method used for conventional single-gate structure ISFETs, in which the IPG electrode is grounded and the drain current is measured while scanning the top reference electrode voltage. In contrast, a dual-gate (DG) mode sensing is possible only in TFTs with a double-gate structure, wherein the top reference electrode is grounded and the drain current is measured while scanning the IPG voltage [13–16]. Moreover, a ceramic-plug junction (Horiba 2086A-06 T) filled with KCl and AgCl saturated solutions was used as the reference electrode for pH sensing. In this work, the pH sensing characteristics of the SG mode were measured by biasing the top reference electrode using the grounded IPG. In contrast, the pH sensing characteristics of the DG mode were investigated by scanning the voltage at the IPG using the grounded top reference electrode [14,14,15,16]. All measurement method to investigate the characteristics of the transducer and pH sensor were shown in supplementary data (Fig. S1). Fig. 1 shows the schematic of the DG mode measurement using the fabricated IPG a-IGZO TFT, as a transducer, and the EG, as a detector. We defined the gate voltage as the reference voltage (V_R) when the reference current (I_R) is 1 nA in the SG and DG modes. The hysteresis and drift characteristics were measured to evaluate the reliability and stability of the fabricated IPG TFT pH sensor. The hysteresis characteristics were determined from the V_R difference between the initial pH 7 and final pH 7 solutions using a pH loop of 7–10–7–4–7, and the electrical characteristics were measured by monitoring the V_R of each pH solution five times at intervals of 2 min. The drift characteristics were defined as the ratio between the initial and final V_R values when the pH 7 solution is placed on the sensing membrane for 8 h. The all electrical characteristics were measured using an Agilent 4156B semiconductor parameter analyzer and a reference electrode; the measurement was done in a dark box to avoid noise from the outside environment.

3. Results and discussion

In the ISFET, the electrical contact with the electrolyte is achieved using the reference electrode. Thus, the detection of pH of an electrolyte or biological element is typically performed by sweeping only the top gate reference electrode voltage; this is defined as a single gate sensing scheme. Therefore, the threshold voltage shift in the conventional SG mode depends largely on the change in the surface potential, and the sensing capability has a Nernst limit of 59 mv/pH at room temperature as per the Nernstian equation [11,12]. For biosensors that need to detect bio-materials with very small potentials, such as living cells, antigen-antibodies, and DNA, the low-sensitivity limitations are a major obstacle to the development of bio-sensors. Therefore, attempts have been made to overcome this problem by developing and studying various aspects of the material and structure of the sensing membrane [28–30]. In contrast, there have been reports on increasing the sensitivity of the sensors without requiring additional processes or circuit configuration using the capacitive coupling effects between the top- and bottom-gate oxides of the transistor based on SOI and a-IGZO [13–16]. The relationship between the capacitive coupling effects and the sensitivity of a transistor with a double-gate structure can be expressed using the following equation [15,16].